U. Oregon

A NEW CONCEPT OF ELECTRICAL IMPEDANCE PLETHYSMOGRAPHY*

Preliminary Report

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Status Report for period ending 5/31/65

Electrical impedance plethysmography has been used as a tool for indirectly measuring blood pulsatile volume changes in many different living tissues. The accepted theory[1] is that there is less electrical resistance in a segment of tissue during systole because of the momentary increased volume (due to influx of blood) of that tissue. The volume conductor formula: $R = \rho \frac{l^2}{V}$, where R is resistance in ohms, ρ is the specific resistivity of the conductor, l is the length in centimeters, and A is the cross-sectional area, provides the basis for this theory. Certain paradoxes have been noted, however. The signal has occasionally appeared to be inverted[2], and to have lost amplitude for no apparent reason. These paradoxes are now explainable on a more logical basis than have heretofore been given.

From our work in the development of the Ocular Electrical Impedance Plethysmograph for NASA (Contract NAS 2-1631) these paradoxes often occurred, especially with the earlier rigid assemblies of contact electrodes. It was, therefore, agreed between NASA Ames Research Center and the University of Oregon Medical School that a follow-on Grant for the purpose of "biological calibration" of the ocular impedance plethysmograph should be let.

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The calibration procedures soon led us to the enucleated, blood-perfused bovine eye, which allowed us to control the environmental and physiologic parameters. It has been shown that freshly enucleated bovine eyes may be "brought to life", as evidenced by the return of the ERG (electro-retinogram) when the arterial system is perfused with heparinized bovine blood[12]. Using this technic, with the four-terminal electrode system mounted on the bare sclera (both in pads, as well as by sutures), we made the following observations:

1. Increasing the blood-volume in the eye increased the electrical resistance,
2. Flushing the vascular tree with isotonic saline decreased the electrical resistance,
3. Pressure on the electrodes, either direct or indirect (by stretching the sclera with increased intra-ocular pressure), decreased the electrical resistance.

Bovine and human blood were then measured, both in whole as well as in centrifuged states, and the following values were obtained. Isotonic saline was found to be several times more conductive (average 73.5 Ω) than blood (average 243 Ω). It is assumed that tissue fluids will be found to be at nearly the impedance of isotonic saline, but even if greater, certainly not as great as that of whole blood. Thus it appears that the "normal" electrical impedance plethysmograph should be inverted rather than upright, if the convention of representing the upward deviation as representing increased conductivity with

blood-filling is still to be followed. What is more important, judgment and evaluation of signals which contain the algebraic summation of oppositely occurring phenomena (polarity up pressure on electrodes, down with blood-filling) cannot be correct unless both are accurately measured simultaneously, or unless one is held constant. The description of electrical impedance plethysmograph instrumentation to date[1-11], including ours, does not include this needed provision. The following two approaches to the solution of this problem are herewith presented. Both systems appear to have equal merit at this time.

A

Figure 1 presents the various impedances of a given biological tissue sample in diagrammatic form using a four electrode plethysmographic system. Although a capacitive component is not shown in the diagram, the resistive elements are to be taken as reactive impedance, whose reactance involves many sources.

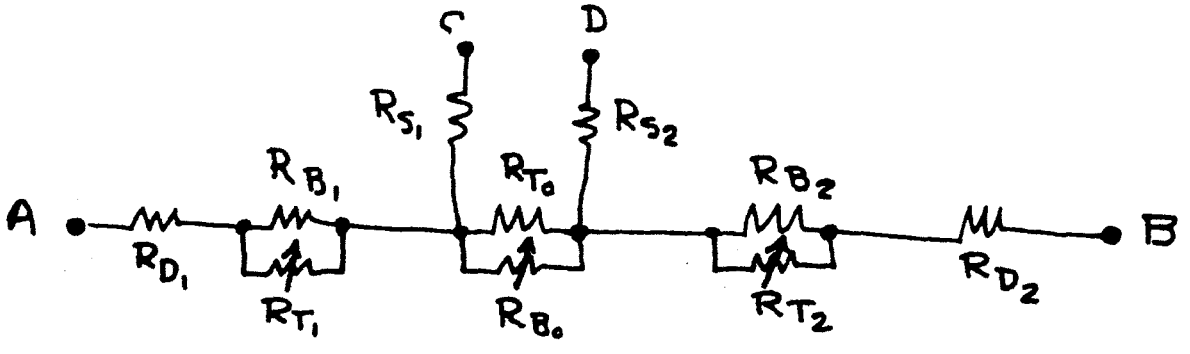

FIGURE 1

Points A and B represent sources of differential or "push-pull" driving current to a biological preparation. R_{D1} and R_{D2} are the impedances at the driving electrode-tissue interface. Points C and D represent voltage or current sensing elements, i.e., preamplifiers or similar sensing devices.

R_{S1} and R_{S2} are the impedances at the Sensing Electrode-Tissue interfaces. The R_{T1B1} and R_{T2B2} symbols represent the relatively stable tissue impedances and the active blood impedances between the driving and sensing electrode interfaces. R_{To} and R_{Bo} are the impedance systems which are to be measured according to ohm's law, IR=E.

Accepting the basic premise, that the sum of the IR drops (E) in any given closed circuit is equal to the source E, it is apparent that when A and B are connected to a common source E, any of the impedances if they change (excepting R_{S1} and R_{S2}) can effect the current flow and therefore the developed E at any point in direct proportion to their change. It should be noted that R_{D1} and R_{D2}, although not diagrammed as being active, are indeed active at the interface contact area. Contact impedance is, therefore, a function of the force which secures the electrodes. When active measurements are made sensitive to very small changes, the contact impedance causes a variance in the current flow through R_{Bo} and R_{To}, which can be a significant error.

One more premise must be applied at this point. In any electrolyte-metal interface, half cell reactions (polarization), will be present with current flow. This EMF, if affected under DC could present an error. If, however, AC is used and relatively unreactive electrodes are used (Ag-AgCl) the effects can be reduced to a level which can be tolerated. This principle is well accepted.

When points C and D are connected to any input, current flows, and its path can become involved. It is primarily in shunt to R_{To} and R_{Bo}. The less current that is allowed to flow (by use of very high input impedance sensing devices) the less significant it becomes as an active part of the

desired signal, the IR drop across R_{To} and R_{Bo}. The obvious advantages of differential input (points C and D) in the rejection of common mode signals is applicable to this situation where changes of less than 0.1 Ω are expected for the R_{To} and R_{Bo} combination.

It is apparent that to observe the E developed by current flow through R_{To}, R_{Bo} for a particular mitigation of this ratio, all error currents which are subject to changes in the other impedance elements must be either eliminated or algebraically removed. Since it is problematical that the interface impedances can be removed by mechanics and the R_D, $R_{T1,2}$ and $R_{B1,2}$ impedances cannot be so removed it is assumed that electronic methods must be developed.

The Impedance Plethysmograph developed under NASA Contract NAS 2-1631, [11] although subject to the errors just described and which have recently been well established, lends itself with modification to a time-sharing error sampling system. (Figure 2.) This error can be applied to the Plethysmographic signal to produce a true impedance measurement coincident with blood flow.

The following general steps will be necessary in the development of such a system. (Figures 3 and 4.)

1. Addition of a balanced differential drive system.

2. During the cyclic period of the driving current in the present instrument only one half of the cycle is demodulated to give the output signal, the other half is available for use of the error signal. A gating system will be developed to shunt current alternately, every other cycle during the non-used time portion through one set of the Driving Electrode-Tissue interface --

and the Sensing Electrode-Tissue interface only, by-passing R_{Bo}-R_{To}, by grounding the appropriate Sensing Electrode at point C or D. The current will be monitored at A or B, appropriately. The opposite driving source will be gated open to exclude parallel current flow.

3. The error signal developed in part 2, will then be demodulated using conventional techniques, similar to the system used in the present instrument.

4. The error signals can then be added algebraically in an appropriate manner to the output of the instrument or applied to a current compensating system. This area will have to be considered and tested to find the most stable and reliable technique.

Theory of proposed circuit:

The signal which drives one side of the eye is inverted and applied to the opposite side of the eye, producing a balanced driving current around ground.

The output signal, developed across R_{TB} is presented to two amplifiers in a differential configuration. The outputs from these amplifiers are added to produce a single ended signal which is then nulled with an opposite polarity signal from the driver, calibrated to readout in ohms. This reading is representative of R_{TB}. Demodulation is performed as before.

Since demodulation only occurs during one half of the driving cycle, the unused portion is the period in which the error signal is to be developed.

The negative going portion of the differential driving voltage is applied through steering diodes to a bistable multivibrator, which acts as a count back circuit. The output is again differentiated and applied to a monostable multivibrator which alternately gates the preamp inputs to ground, and gates the opposite output driver to a high impedance or "open" condition. Current flows, during this cycle, through R_D, R_T and R_S to ground. The signal developed at A or B respectively is proportional to the impedance.

These error signals are then amplified and appropriately inverted to produce a positive d.c. voltage proportional to the error. This voltage is gated so that it is held during the other portion of the cycle which represents impedance plethysmograph current flow.

The error voltages are then algebraically added to the signal from the demodulated output of the apparent plethysmographic voltage, which includes the variation of IR_{TD} due to the driving electrode impedances change. The resultant signal after correction is the true plethysmographic signal.

B

PREMISE

A second system for correcting the errors which are generated by the driving electrode tissue interface impedance changes is based on constant current. If there is constant current flowing through the entire impedance string, any variations in the individual series impedance components will not be reflected to R_{TB}. Inasmuch as the signal which is demodulated and represents

the Impedance Plethysmographic signal is a function of I x R_{ToBo}, it is recognized as a voltage drop, E. Therefore, if I is constant, and E is the signal, the only variable is R_{ToBo}. And, since it is obvious the $R_{D1,2}$, $R_{T1,2}$ can also be variable, it continues that with constant current only the individual IR drops change. Constant current demands a power driving source that can compensate for the varying IR=E by supplying a variable E to correct.

CIRCUIT THEORY

Each driver in a balanced driving system will be controlled by a variable signal source.

An emitter coupled differential input amplifier (current control amplifier) will be used to control the driving signal level. The emitter follower input side will be adjustable and referenced to the shaper signal. This will be the current level adjustment. The amplifier side will be referenced to the output of an operational amplifier. This amplifier's input is across a precision sampling resistor in series with the drive current, and the eye. Any changes in current through the sampling resistor and hence the entire impedance string are reflected as voltages to the current control amplifier. The signal is in phase with the current level signal. At the collector output of the current control amplifier the signal is the algebraic difference and proportional to the ratio between the two inputs. The voltage across the eye is then a function of this difference and proportional to changes in I through the sampling resistor.

If the current increases through the eye, the IR drop across the sampling resistor will increase, and a greater signal will appear at the current control amplifier. This will tend to decrease the difference at the output, hence the drive signal and therefore the current. This method also works in reverse.

The problems with this concept is in electronics which can recognize and correct changes in the order of 10μ volts (10×10^{-6} volts) and less. This is based on a driving current of 100μ amperes and 0.1Ω changes. The problems of stability in components are much greater at these levels. However, much work is being done at these levels and the development of components and systems in producing better instruments continuously.

DISCUSSION

Additional experiments on the human finger and over the human carotid artery also appear to confirm our concept. The carotid experiment is easily performed and is the more dramatic. The four-terminal plethysmograph electrodes are held alternately lightly and then with deep pressure over the carotid artery area on the neck. With light pressure the signal is upright, with deep pressure, it is inverted. Thus it appears that when the smaller increments of pressure by systolic blood-filling are "swamped-out" by the total average external pressure on the electrodes over the carotid artery, the true impedance plethysmograph then appears. The same phenomenon occurs on the finger, but with less amplitude.

Also, with the blood-perfused bovine eye, utilizing micrometer adjusted pressure on the electrodes, monitored by a Statham strain-gage, a cross-over point was found, where infusion of blood caused the signal to invert, depending upon whether the electrode pressure increased conductance more or less than perfused blood decreased conductance. These, and other similar events have been recorded, and will be published at a later date.

SUMMARY

A new concept for the theory of operation of the well-known electrical impedance plethysmograph is presented. The volume-conductor theory, while undoubtedly playing some role, does not appear to be as important as electrode pressure in the explanation of the apparent increased conductance of electrical current through blood-filled tissues. A new type of electrical impedance plethysmograph, which will eliminate this error, is proposed.

ACKNOWLEDGEMENT

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MAY 1965

THE PRESENT CIRCUIT:

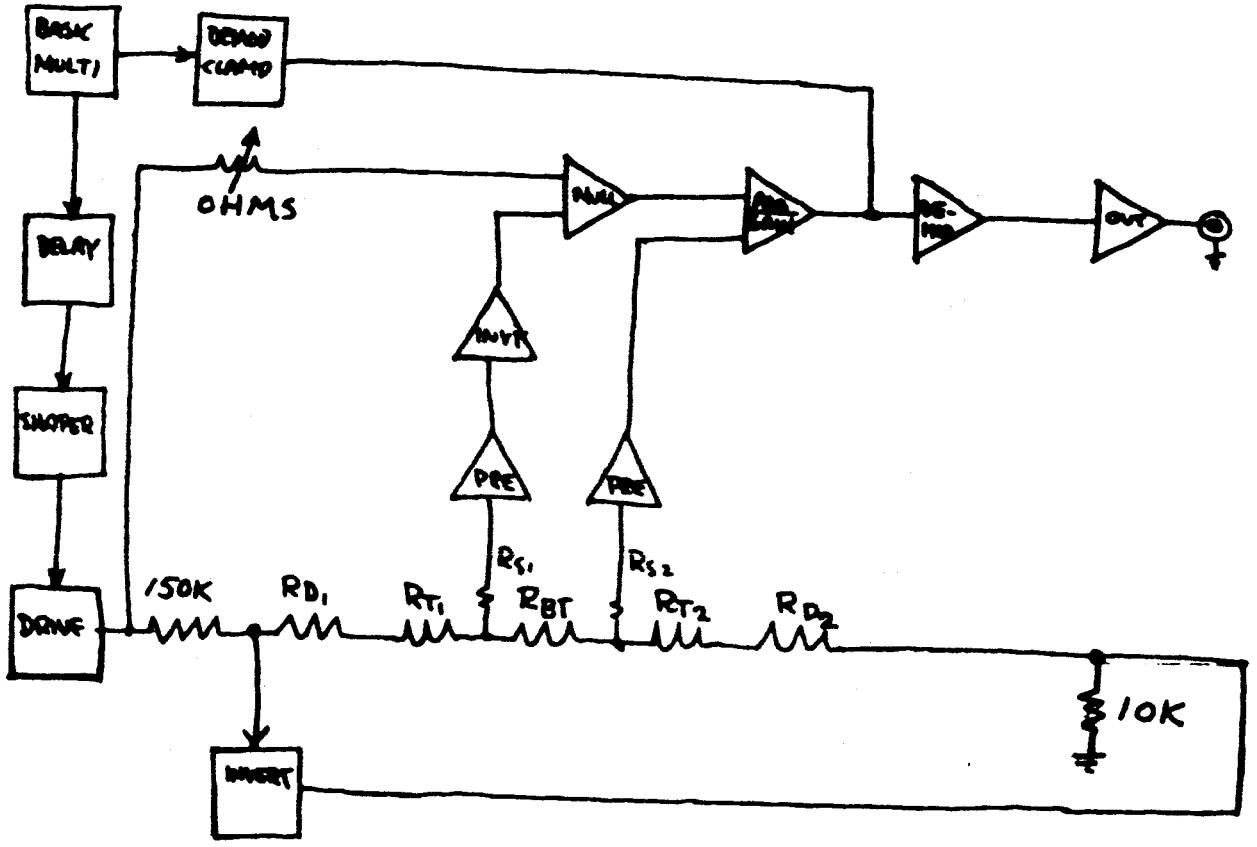

FIGURE 2

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PROPOSED REVISIONS TO PRESENT CIRCUIT

▬ MODIFICATION AND ADDITIONS

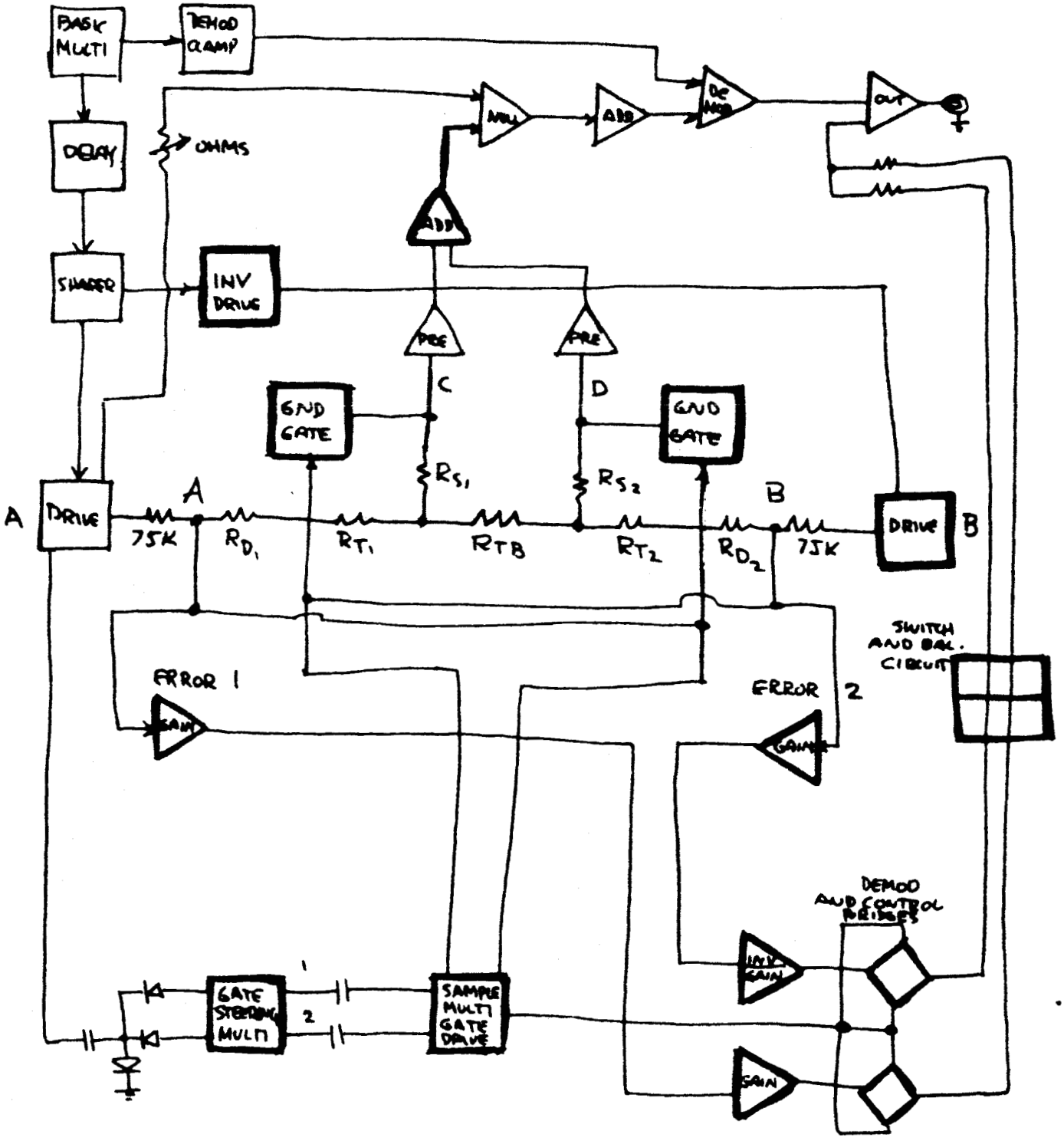

FIGURE 3

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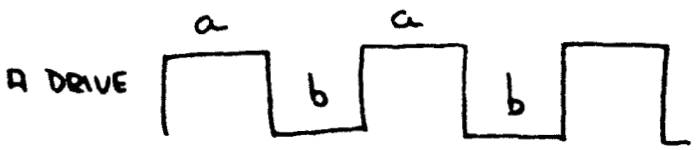
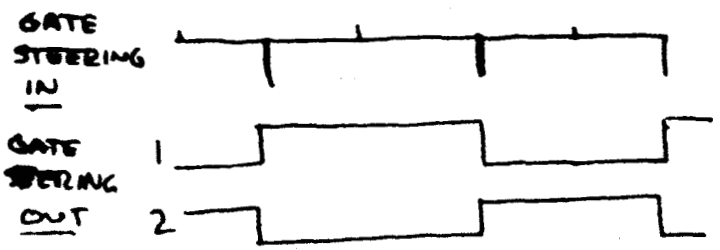
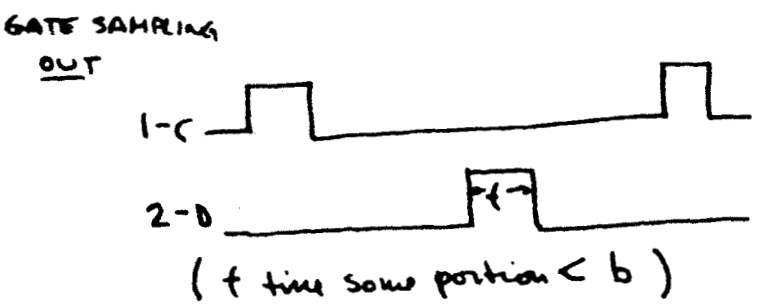
PROPOSED CIRCUIT WAVEFORMS

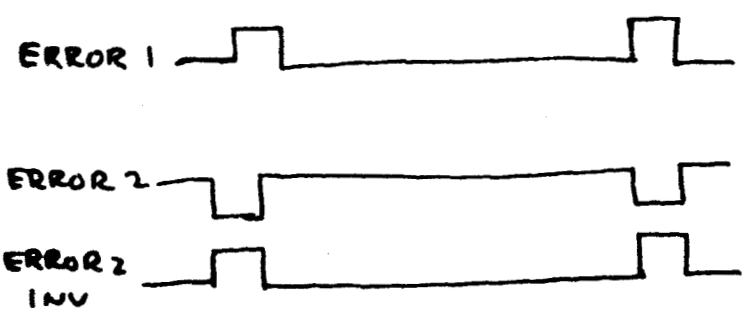


a portion demod for Z plethysmograph signal





(t time some portion < b)



where amplitude of ERROR signal is $\cong$ to IR_{D_1}, IR_{T_1}, IR_{S_1} or IR_{D_2}, IR_{T_2}, IR_{S_2}

THE ERROR VOLTAGES ARE RECTIFIED AND APPLIED AS DC POTENTIALS TO AN ALGEBRAIC SUMMING CKT.

FIGURE 4

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CONSTANT CURRENT SYSTEM

■ = DENOTE ADDITION

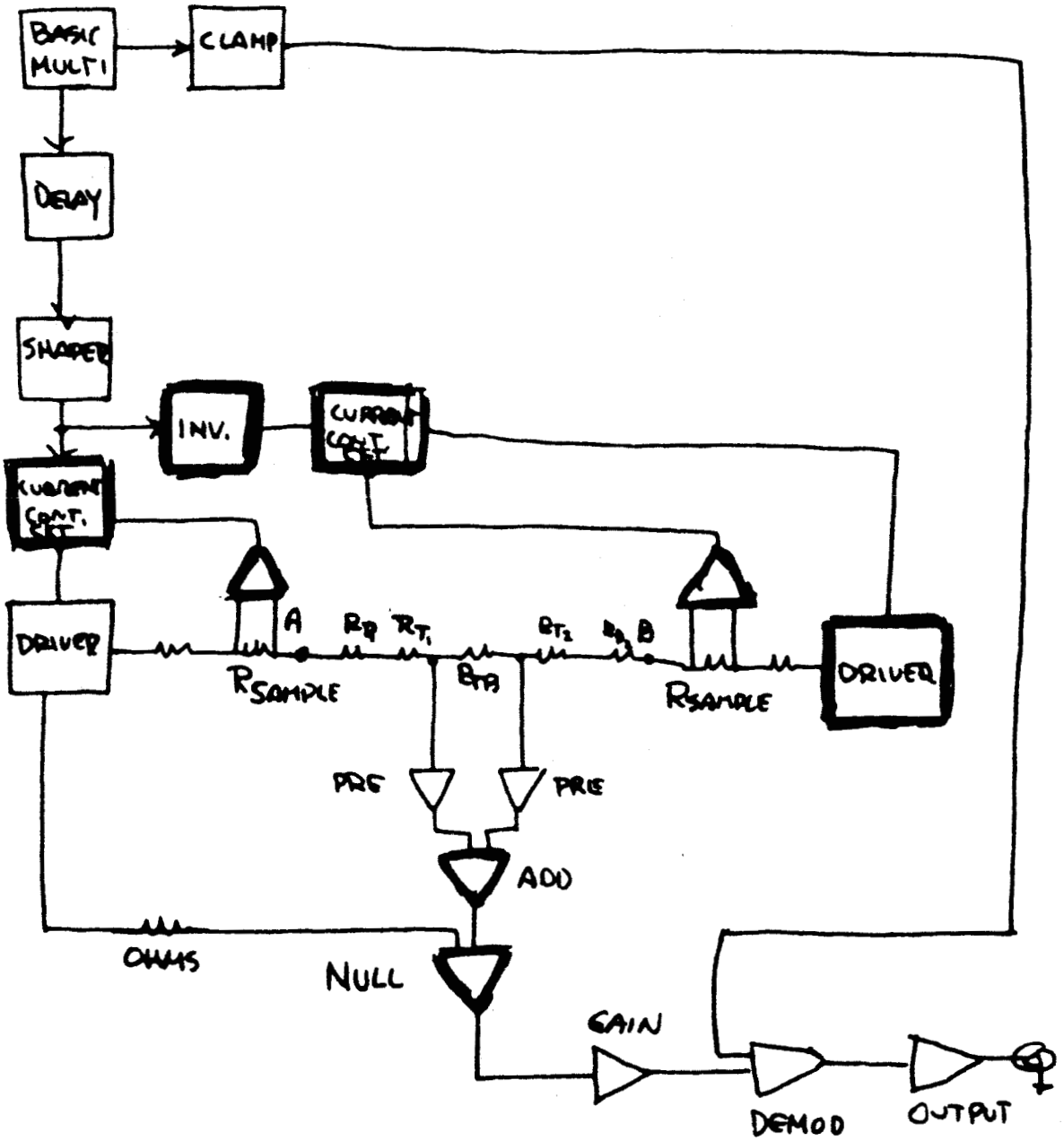

FIGURE 5

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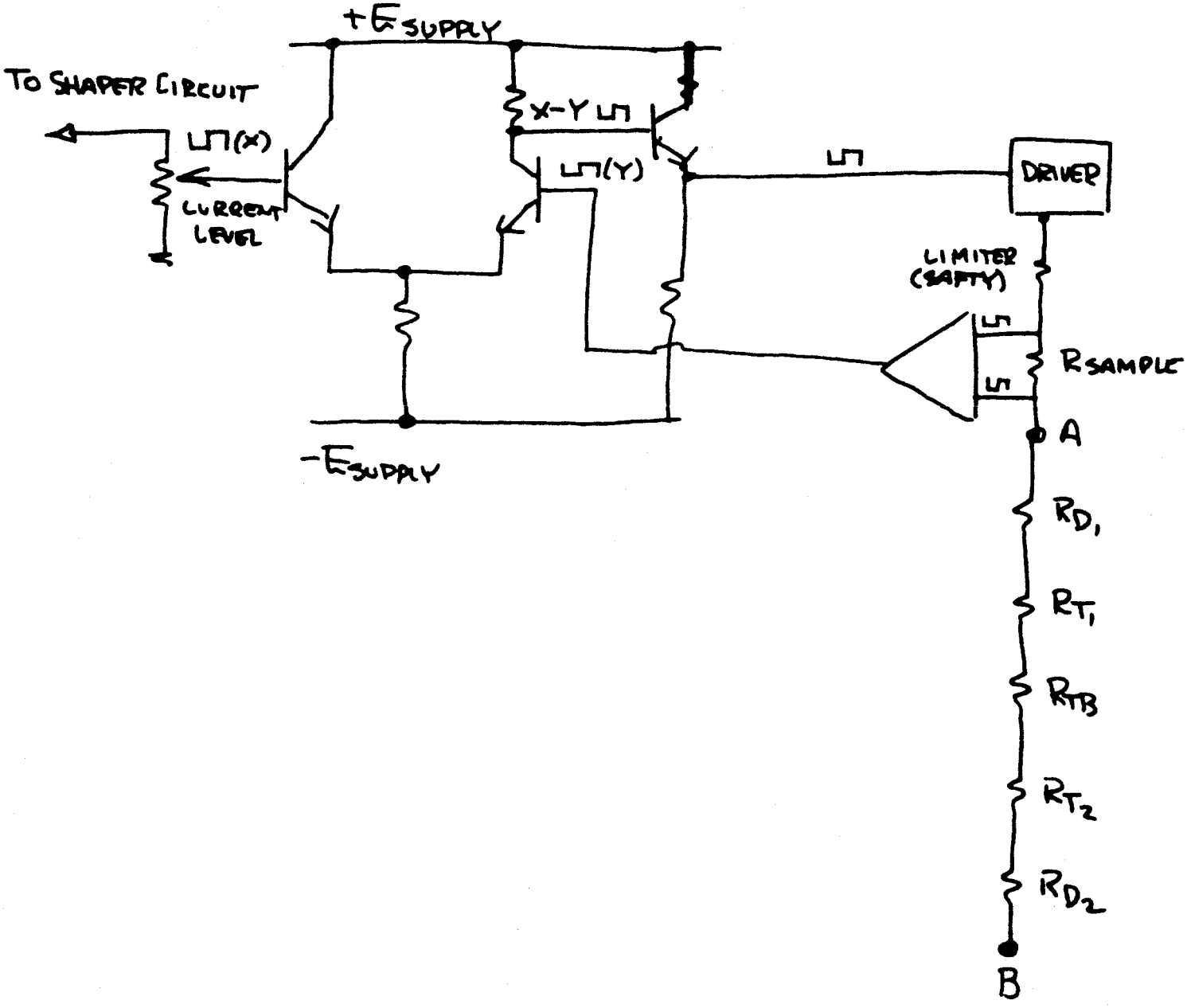
CURRENT CONTROL AMPLIFIER



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